THIN-FILM INTRACORTICAL RECORDING MICROELECTRODES

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Thin-Film Intracortical Recording Microelectrodes

Summary

During the past quarter, we have continued to explore the long-term recording characteristics of these probes in-vivo. One 8-shank probe having both large tip sites and standard center-shank-mounted sites is still recording well from guinea pig auditory cortex after 249 days. We have also explored the use of rough gold plating and found it effective in reducing site impedances as well as in providing increased adhesion for electrochemically-deposited conducting site polymers such as polypyrrole doped with polystyenensulfonate (PPy/PSS).

A new high-density connector has been built using a polyimide-based flex circuit rather than the traditional rigid FR-4 board. The flex circuit is double-sided which permits two dual-row Omnetics connectors to be used to achieve 32 channels in a package measuring 12mm tall x 9mm wide x 5mm thick. Implants using these packages will begin during the coming quarter. Also during the past term, we have begun to re-examine the possibility of reducing probe shank width to dimensions below 10µm. This should result in significantly-reduced tissue damage on insertion and permit lateral recording between shanks within the recording field of single cells. We will explore the fundamental limits on shank width as well as the use of alternative technologies such as silicon-on-insulator to achieve greater compatibility with foundry fabrication of the probes.

A complete graphical user interface for the non-multiplexed 64-site 8-channel PIA-2B probe has been developed, and work with this probe is proceeding in collaboration with Gyorgy Buzsaki's laboratory at Rutgers University. The probe has reduced recording artifacts and been used to drive ten-foot cables without a headstage. Results at Rutgers using the passive version of a companion 96-site probe in rat hippocampus were reported in the July 17 issue of *Nature*. We have also designed a version of the 96-site probe using op-amp-based buffer amplifiers and look forward to using these devices during the coming term.

Several versions of multiplexed 64-site 8-channel active probes (PIA-2 and PIA-3) have been designed and are now in fabrication. The probes feature different amplifier designs and a variety of test modes. In addition, we have designed a probe for use in a completely wireless recording system. The probe features an on-chip gain of 31.5dB, a bandwidth from 1Hz to 10kHz, a power dissipation of 0.32mW, a layout area of 0.24mm^2, and a swing of ±1V from a ±1.5V supply. A platform-mounted telemetry interface for the probe has been designed and submitted to MOSIS for fabrication. The interface dissipates less than 1mW in a layout area of 2.2mm x 2.2mm. We hope to demonstrate completely-implantable wireless neural recording before the end of the year.

Thin-Film Intracortical Recording Microelectrodes

1. Introduction

The goal of this program is the realization of batch-fabricated recording electrode arrays capable of accurately sampling single-unit neural activity throughout of volume of cortical tissue on a chronic basis. Such arrays will constitute an important advance in instrumentation for the study of information processing in neural structures and should be valuable for a number of next-generation closed-loop neural prostheses, where stimuli must be conditioned on the response of the physiological system.

The approach taken in this research involves the use of solid-state process technology to realize probes in which a precisely-etched silicon substrate supports an array of thin-film conductors insulated above and below by deposited dielectrics. Openings in the dielectrics, produced using photolithography, form recording sites which permit recording from single neurons on a highly-selective basis. The fabrication processes for both passive and active (containing signal-processing circuitry) probe structures have been reported in the past along with scaling limits and the results of numerous acute experiments using passive probes in animals. In moving to chronic implant applications, the major problems are associated with the preserving the viability of the sites in-vivo (preventing tissue encapsulation of the sites) and with the probe output leads, both in terms of their number and their insulation. The probe must float in the tissue with minimal tethering forces, limiting the number of leads to a few at most. The encapsulation of these leads must offer adequate protection for the megohm impedance levels of the sites while maintaining mechanical lead flexibility.

Our solution to the lead problem has involved two steps. The first has been to embed circuitry in the probe substrate to amplify and buffer the signals and to multiplex them onto a common output line. Using this approach, signal levels are increased by factors of over 100, impedance levels are reduced by three to four orders of magnitude, and the probe requires only a few leads for operation, independent of the number of recording sites. A high-yield merged process permitting the integration of CMOS circuitry on the probe has been developed, and this circuitry has been designed and characterized. The second step has involved the development of silicon-based ribbon cables, realized using the same probe technology, to conduct the neural signals to the outside world. These cables have shown significant advantages over discrete leads, both in terms of the ease with which chronic implants can be assembled and in terms of the ability of the cables to survive long-term biased soaks in saline. The cables can be built directly into the probes so that they come off of the wafer as a single unit, requiring no joining or bonding operations between them. The cables are also significantly more flexible than previously-used discrete wire interconnects.

This contract calls for the development of active probes for neural recording. A 64-site 8-channel probe with site selection and signal buffering but no multiplexing has been developed (PIA-2B) along with a high-end multiplexed probe that includes gain

(PIA-2). These probes are now being refined and applied to in-vivo applications. Investigations are on-going to better understand site encapsulation, which limits the lifetime of chronic recording structures, and telemetry is being developed to allow the probes to be operated over a wireless link, eliminating the percutaneous connector.

During the past quarter, we have continued to study recording lifetime as a function of site size, placement, and various biocoatings. Active probes (PIA-2B) are now in use and in-vivo data is being gathered. The multiplexed PIA-2/3 probes have been redesigned and a series of these devices having different amplifier configurations are now in fabrication. Layout has also been completed on a wireless interface for the probes with the intent of demonstrating wireless recording by the end of the year. Work in these areas is discussed in the sections below.

2. Passive Probe Development

Chronic Animals:

During the past quarter, we have continued to work on improving the length of time that we can record from our chronic animals before the signal degrades. This work is being done in conjunction with the Center for Neural Communication Technology. We are continuing to record from "MW4". This guinea pig has been implanted into auditory cortex with an eight-shank recording electrode. The electrode has a 1000μm^2 site at the tip and a standard size site (about 100μm^2) 25µm above that (Fig. 1).

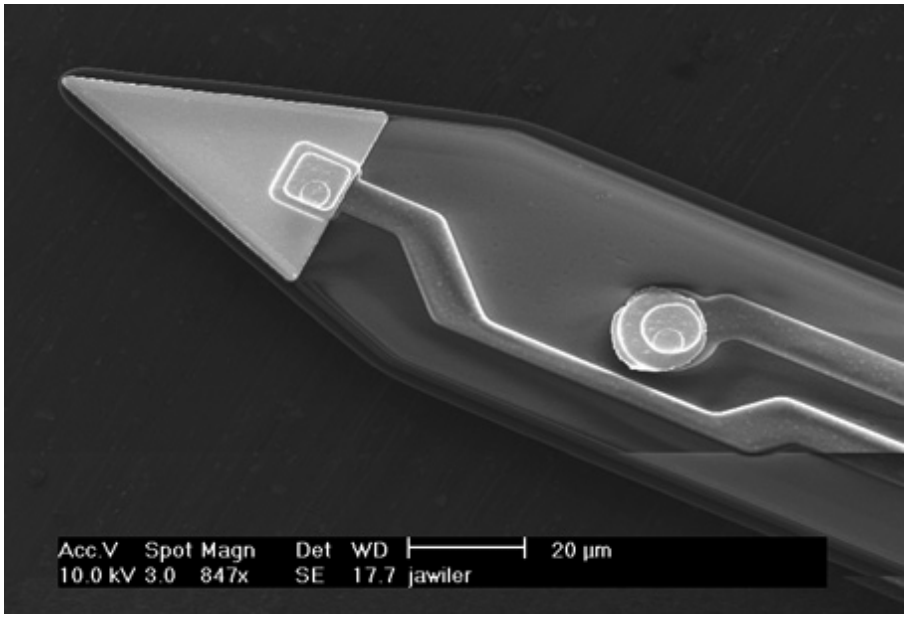

Fig. 1: SEM of a new probe designed to investigate effects of tip site placement on chronic recording lifetime. In addition to a typical recording site in the middle of the shank, the probe has a large site at the tip of the shank.

We have now been recording from MW4 for 249 days. Figure 2 shows the site impedances of this probe over this time period. While there have been minor fluctuations, the electrode has been relatively stable overall. Figure 3 shows a 100msec recording obtained on channel 1. We will continue to follow this animal as long as possible.

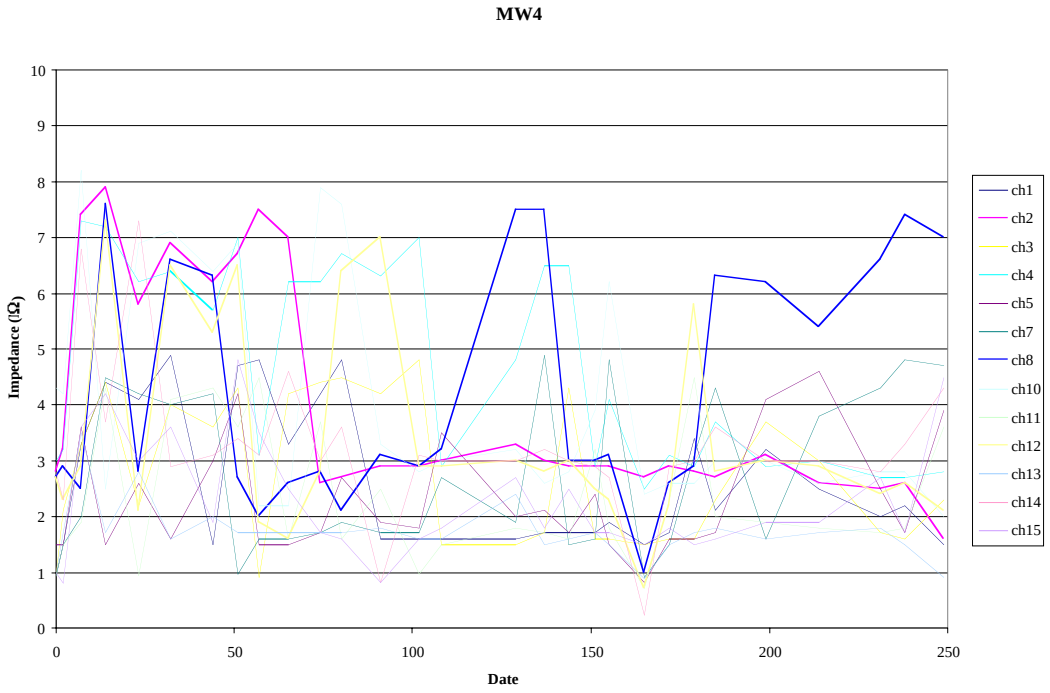

Fig. 2: Impedance measurements for guinea pig "MW4" 249 days post implant.

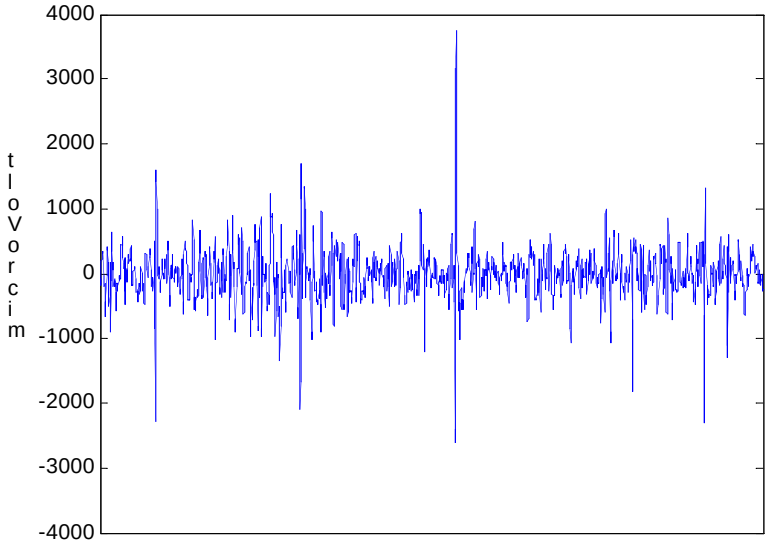

Fig. 3: A 100 msec recording from "MW4" 249 days post implantation.

We have recently implanted another tip site electrode (MW5). It has been implanted for a month, but has only been recording evoked potentials during this time.

3. Biopolymer Coatings

As part of the effort to increase the length of time that we can record from our chronic animals, we have been exploring the use of conductive polymers such as polypyrrole as outer coatings on the recording sites. We are continuing to evaluate the effectiveness of polypyrrole and may try other coatings or combinations of coatings in the coming quarters. One other coating that we reported on in the previous quarterly report was polyethylene glycol (PEG). Guinea pigs "CPEG1" and "CPEG2" were implanted with four-shank sixteen-site probes in auditory cortex. The responses from these animals started to degrade shortly after the beginning of this quarter. The animals were sacrificed and the tissue is being currently being processed for histological examination. We are planning to implant some additional PEG-coated electrodes during the next quarter.

During the past term, we have made considerable progress on biocompatible electrode coatings by confirming experimental methods to deliberately create "fuzzy" gold electrodes by electrochemical deposition. It is well known that surface roughness plays a critical role in site impedance and recording performance. We can significantly increase the effective surface area of the gold itself by electroplating it with high surface roughness. This roughness significantly reduces the impedance of the site and also increases the adhesion of any subsequently deposited conducting polymer films.

An acute 4×4 200μm probe was used in this experiment as shown in Fig. 4. A thin layer of gold was electroplated on every site of the upper two shanks, while the lower two shanks have not been gold plated. We then electrochemically deposited a conducting polymer polypyrrole doped with polystyenensulfonate (PPy/PSS) on 12 of the 16 electrode sites with same amounts used on parallel sites (common depth) and with increasing amounts from left to right. The coated probe was then subject to impedance measurements. The impedance magnitudes of the probe sites were lowered by roughly an order of magnitude after gold electroplating (as shown in Fig. 5), reflecting the increase in the real surface area (shown in Fig. 6).

The adhesion of the polypyrrole film was first tested by rinsing the electrode with water and inserting the probe in agar repeatedly. The coatings stayed attached on the probe. Cyclic voltammetry (CV) was then conducted for 30 cycles. Upon CV cycling from –0.9 to 0.6 V, the conducting polymer undergoes redox reactions that involve diffusion of hydrated ions in and out of the film. The swelling and shrinkage of the film under potential cycling degrades the adhesion between the film and the gold surface and functions as an accelerated adhesion test. A photograph of the probe after 30 CV cycles is shown in Fig. 7. The PPy/PSS films remained attached on all the roughened electrode sites, while 3 out of 6 films detached from the non-roughened electrode sites. This result

indicates a significant improvement of adhesion between polymer coating and electrode substrate by electroplating of gold.

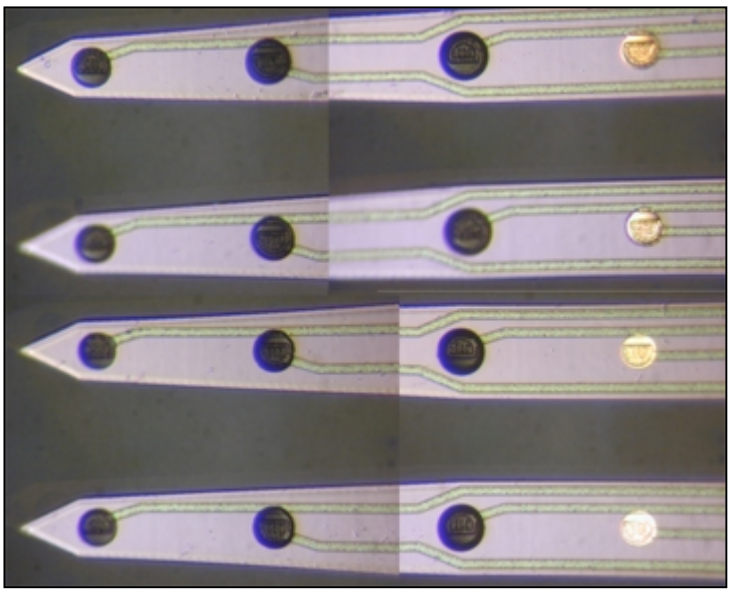

Fig. 4: An acute 4 × 4 probe with sites on 200µm centers. Sites on the upper two shanks were roughened by gold plating, while those on the lower two shanks were not. PPy/PSS was subsequently deposited on the electrode sites with deposition charges of 5, 10 and 20µC from left to right on each shank.

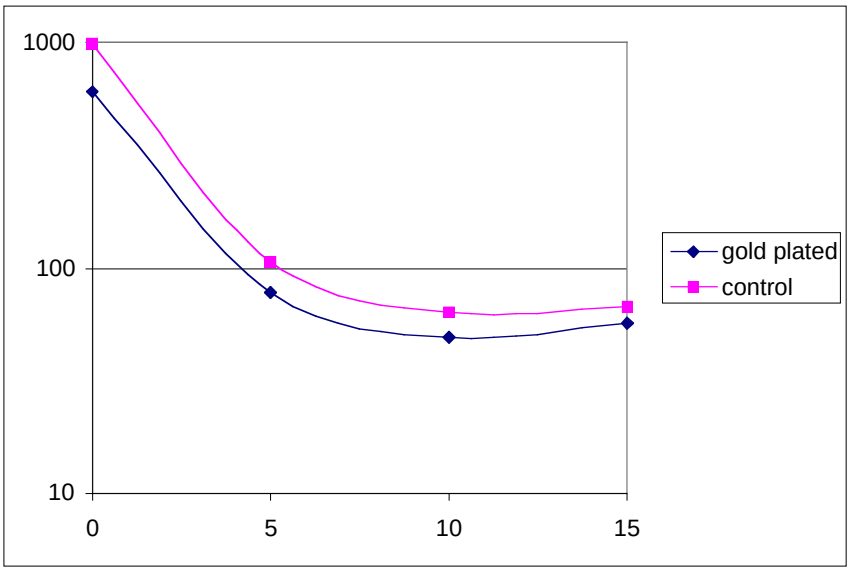

Fig. 5: Impedance magnitude at 1kHz in (kOhm) vs. deposition charge in (µC).

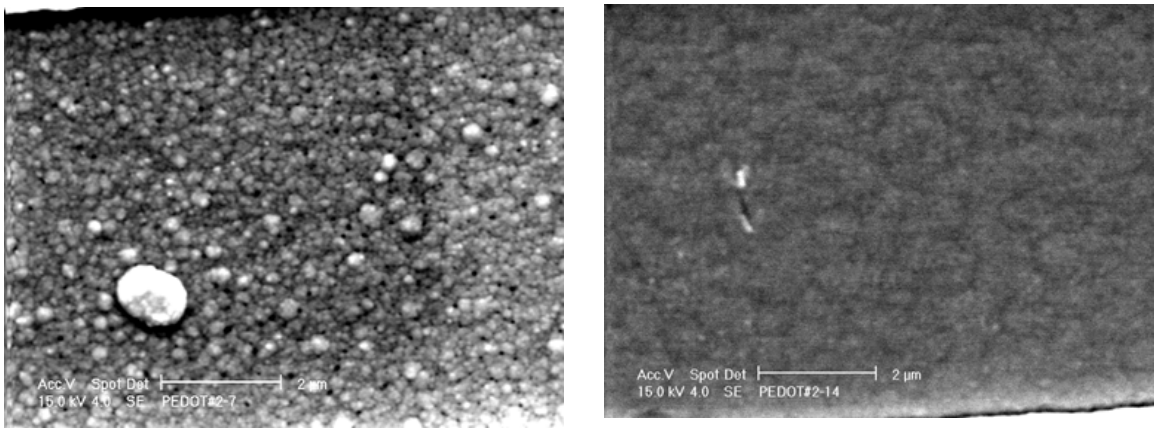

Fig. 6: SEM images of a standard gold electrode site (right) and a site roughened by electrochemical deposition of gold (left).

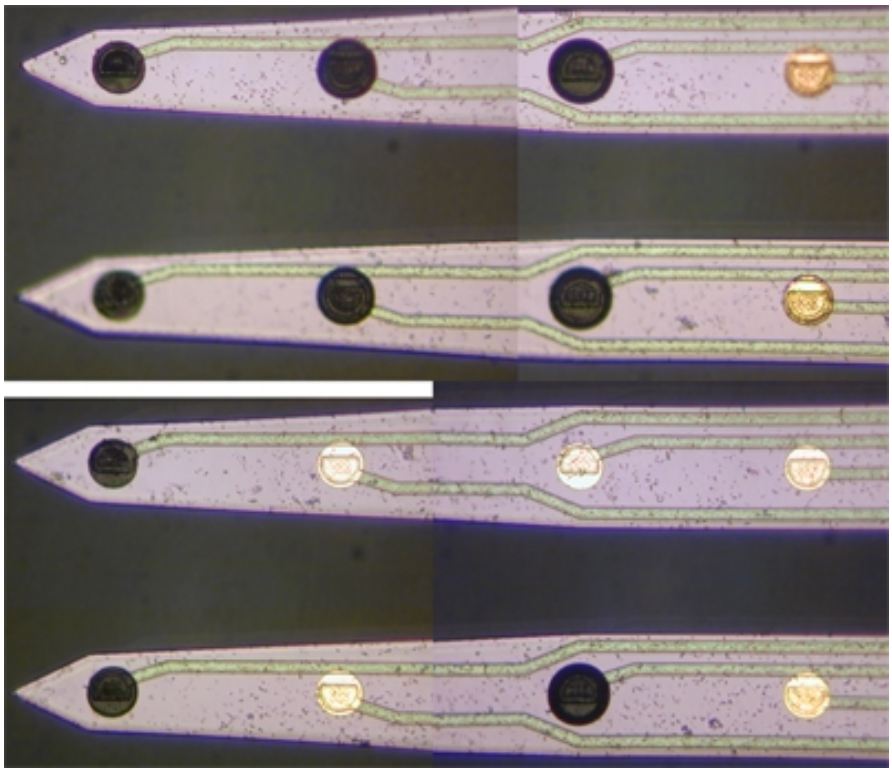

Fig. 7: Optical image of the probe after 30 CV cycles.

Future work in this area will include optimization of the electroplating conditions for higher roughness and the testing of other polymer coatings such as polypyrrole doped with peptides.

4. Packaging Developments

Work to improve chronic assemblies has continued throughout the contract period. While this work is performed primarily under the CNCT, it is relevant to this and other NIH NPP contracts and so is reported here.

The connector to be discussed was developed for several reasons. First, it is intended for "flag" type probes, or those probes that come off at an angle from the integrated cable such as the one designed by Edward Schmidt and Bill Heetderks shown in Fig. 8. In our most common packaging scheme (Fig. 9), the probe is bonded to a PC board that is intended to lie on the skull surface. In this configuration, a flag probe must make a 90-degree bend in order to enter the brain at an angle normal to the surface. Although the silicon cable is very flexible, flexion in this direction will induce torque on the implanted probe. This new assembly alleviates this problem. A second reason for development of this new connector is that it permits 32 channels of output. Several 32 channel chronic "flag" probes have been designed, including two for Daryl Kipke[1] and two chronic chemical delivery probes for Sanford Bledsoe.

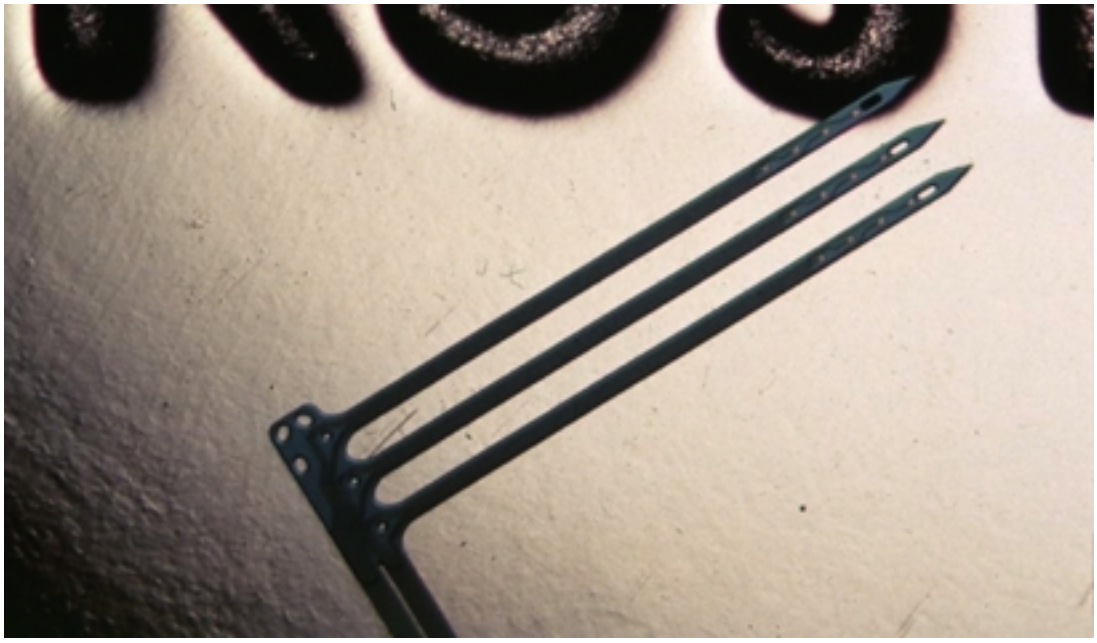

Fig. 8: "Flag" probe designed for Bill Heetderks and Edward Schmidt.

[1] It is noted that Daryl Kipke recently joined the Department of Biomedical Engineering at the University of Michigan. We look forward to working closely with him on these and related programs and feel that he will be a major asset to them.

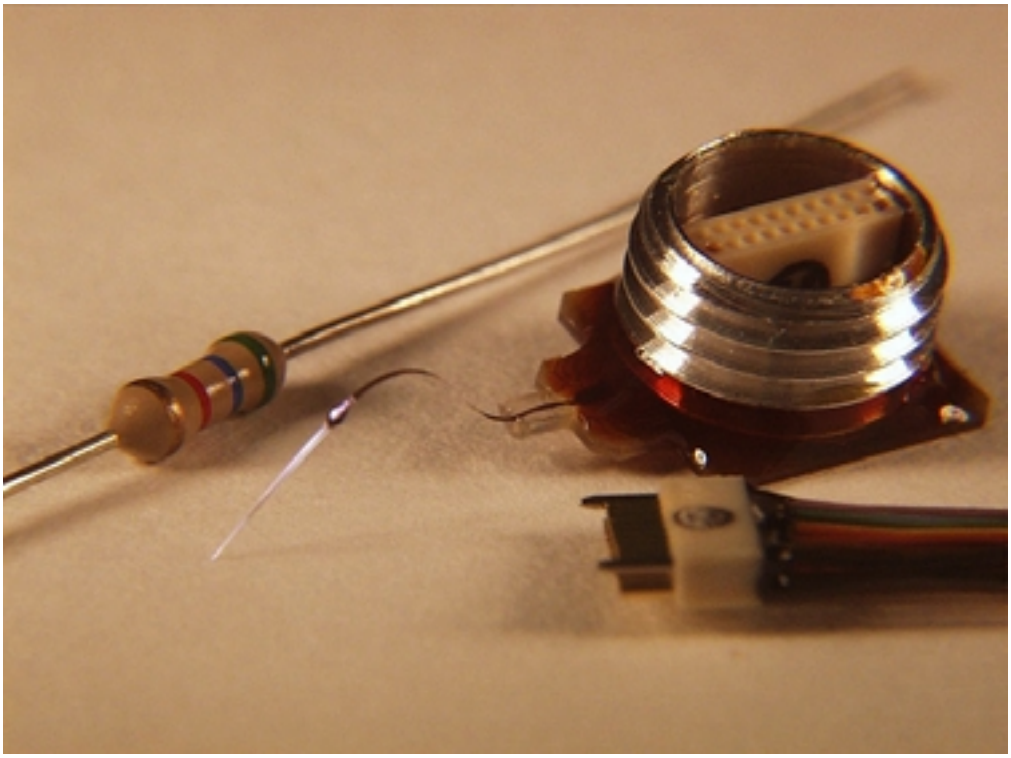

Fig. 9: Standard chronic assembly. If a flag probe is bonded to this assembly, the cable must make a 90 degree turn to permit the probe to enter the brain at a normal angle.

The new assembly is built on a flex circuit, a circuit board that is based on a flexible polyimide substrate rather than a rigid FR-4 board. The flex circuit is double-sided which permits two dual-row Omnetics connectors to be used to achieve 32 channels in a compact package. Assembly takes place as shown in Fig. 10. The probe is mounted and ultrasonically bonded to the flex circuit. Two Omnetics NANO connectors with flex circuit terminations are then soldered to the board. The board is folded in-two, using a fold-over to stabilize its position as well as to form a receptacle that can be filled with epoxy. The device is finally encapsulated with silicone rubber and epoxy.

The completed assembly is about 12mm tall x 9mm wide x 5mm thick. Future iterations may include a receptacle to provide protection between recording sessions and to serve as an anchor within the dental acrylic that holds the assembly on the skull. We will begin implanting these assemblies in the coming quarter.

5. Scaled Recording Probe Development

In the past, we have reported the development of probes having shank widths as narrow as 5µm using an RIE silicon field etch just prior to probe release to cut vertically through the shank-defining boron diffusion and allow probes to be formed with aspect ratios (thickness/width) of one or greater. These probes were not significantly used in-

vivo in the past but promised to allow significantly reduced tissue damage and, perhaps, better chronic recording performance. We have returned to the area of size reduction on these probes in order to explore more fully a number of aspects of their design and use. First, we want to explore the size limits on these probes in the range below shank widths of 10µm (compared with 50µm or more for our "standard" structures). At sizes below 10µm this will require reductions in lithographic feature size, but we have recently fabricated probes having conductor pitches (line plus space) of 3µm and can likely go somewhat lower in-house. In fact, using electron-beam lithography, sub-micron conductor widths are certainly possible. We do not know the penetration characteristics of such sharp tips through pia arachnoid nor hence the adequacy of probe strength at these sizes, but it should be possible to realize highly compliant probes that are quite able to penetrate brain tissue. Since the sites on probe only a few microns wide will certainly hang over the edge of the shank, the recording field should be more nearly hemispherical and may be better in terms of life. Answers to all of these questions should be obtained through the planned study. In addition, with such narrow shanks, reduced lateral shank spacings should be possible, making it equally possible to record laterally as well as vertically from the same unit. For example, on our current probes, with a shank width of 50µm a 200µm lateral shank spacing has normally been used (25% of the lateral space is occupied by probe). With a 5µm shank, lateral spacings of 20-30µm should be possible, well within the recording field of a typical cell. This should be valuable not only in exploring cellular potential fields but also in investigating long-term encapsulation of sites in-vivo. Finally, we will explore as part of this work the use of alternatives to the boron diffusion in exploring narrow probe shanks to make the probes more compatible with outside circuit foundry fabrication. Silicon-on-insulator (SOI) is one such process option. Results in this area will be discussed in future reports.

6. Development of a 64-Site Eight-Channel Non-Multiplexed Recording Probe (PIA-2B)

As reported in past quarters, a front-end-selected and buffered recording probe with self-test features has been fabricated and tested *in-vivo*. The design and architecture of this probe have already been reported, and all digital and analog circuit blocks on the probe function as designed.

To facilitate the use of the probes both here at Michigan and by external investigators, a fully graphical user interface has been developed. This interface was written in Labview for a pentium-based PC, and allows the user to select probe recording and test modes, and to select all possible site selection patterns graphically. The interface system utilizes a National Instruments AT-MIO-16D data acquisition board. The software is easily portable and has been tested on several platforms, operating systems, and different National Instruments DAQ cards. Screen captures of the control panel of the interface system are given in Figs. 10, 11, and 12.

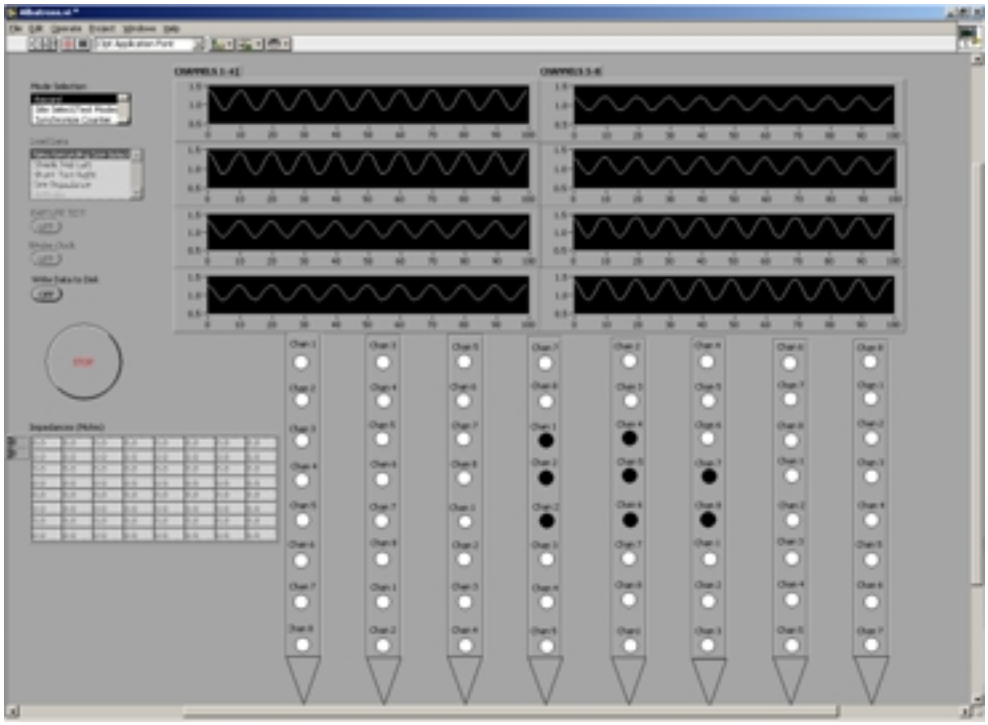

Fig. 10: Screen capture of the control panel of the graphical user interface for front-end-selected probe. A pattern of eight selected sites is shown along with data recorded from a signal injected into saline.

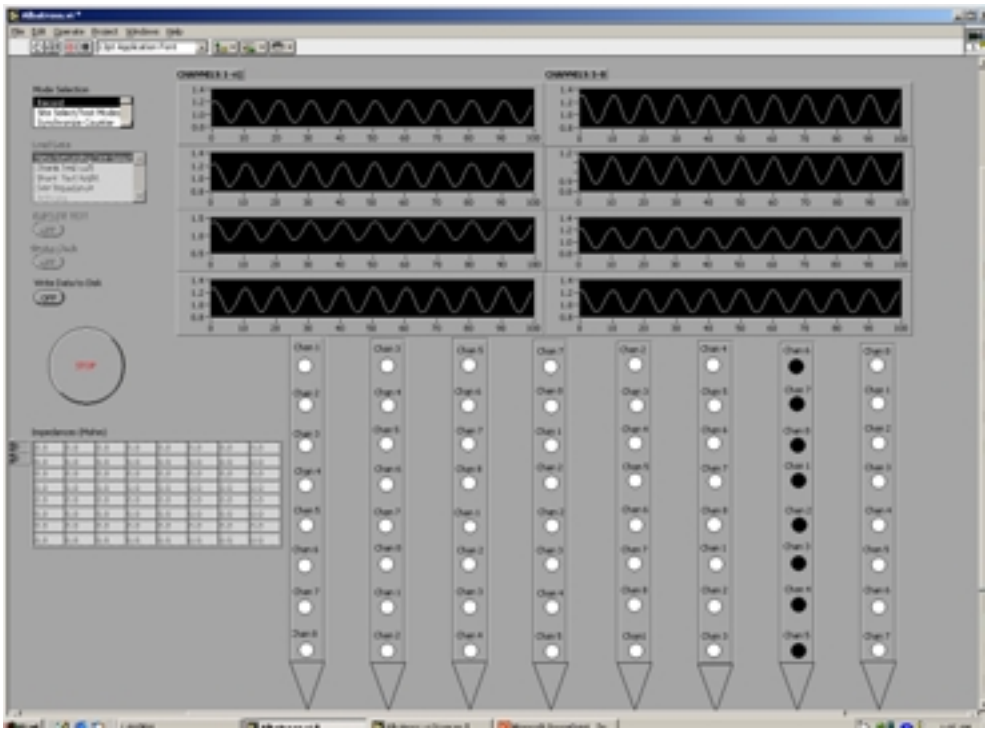

Fig. 11: A different site selection pattern (shank seven.) Notice that the eight output channels exhibit a different pattern than in the previous figure.

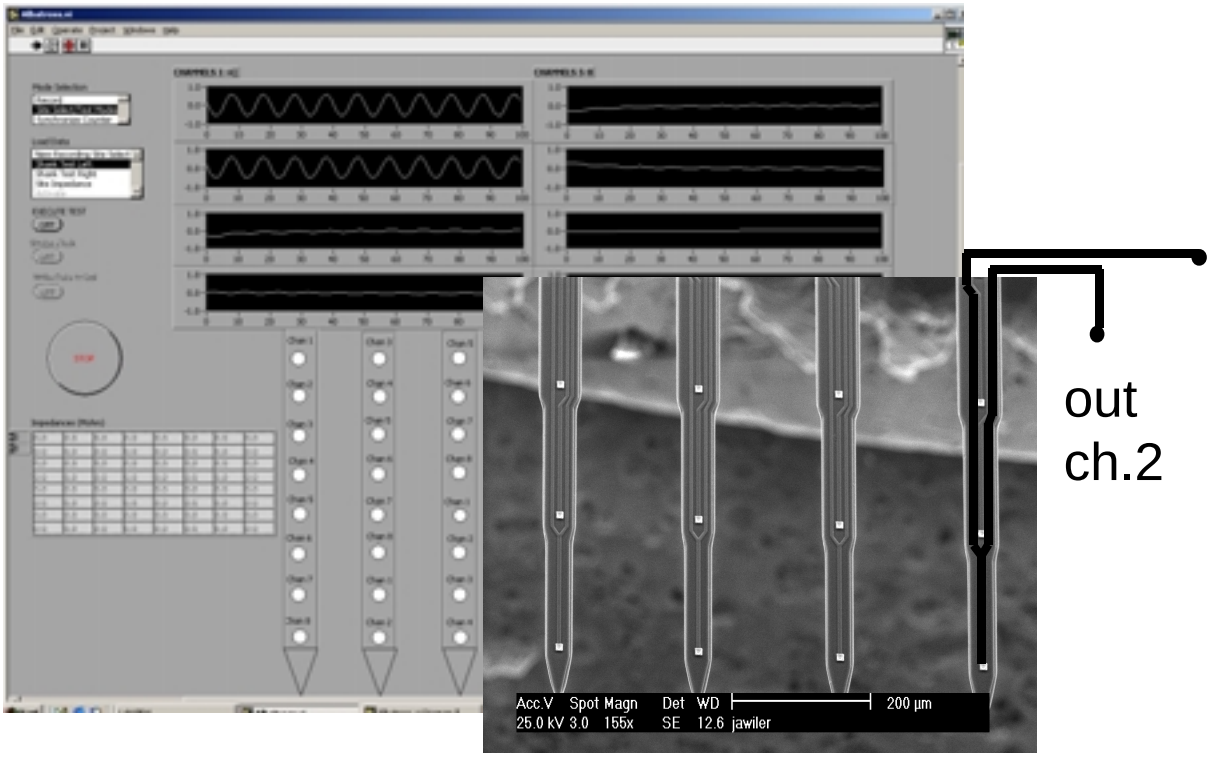

Fig. 12: Control panel during shank test mode. Channels one and two are shorted, while the rest of the channels float. A signal input on channel one is output also on channel 2.

In order to learn more about the functionality of buffered and front-end-selected probes, we have been collaborating with Gyuri Buzsaki's laboratory at Rutgers. Probes with active circuitry represent a potential improvement over passive versions in several experiments being conducted at Rutgers. For single unit recordings in rat hippocampus, it is hoped that the greater number of recording sites per output lead will increase the likelihood of finding an active cell in the correct anatomical region. Such a consideration is especially important where simultaneous intracellular and extracellular recordings are being made, since electrode positioning is critical and difficult. Initial experiments with the front-end-selected probes have been promising. Both field and filtered single unit recordings have been made with the probes, as shown in Figs. 13 and 14, respectively. These recordings were made with the probe driving approximately ten feet of shielded cable, which was connected to an external commercial amplifier stage.

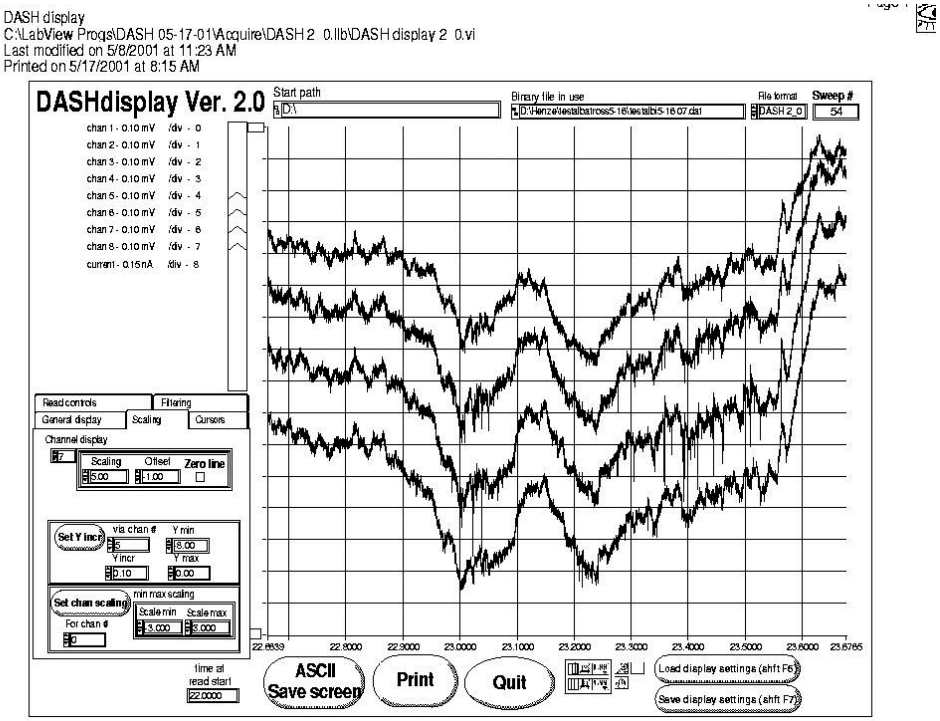

Fig. 13: Wideband (field) recording in rat cortex with a 64 site front-end-selected and buffered probe. As expected, the field activity is highly correlated between sites, which are spaced at 100µm.

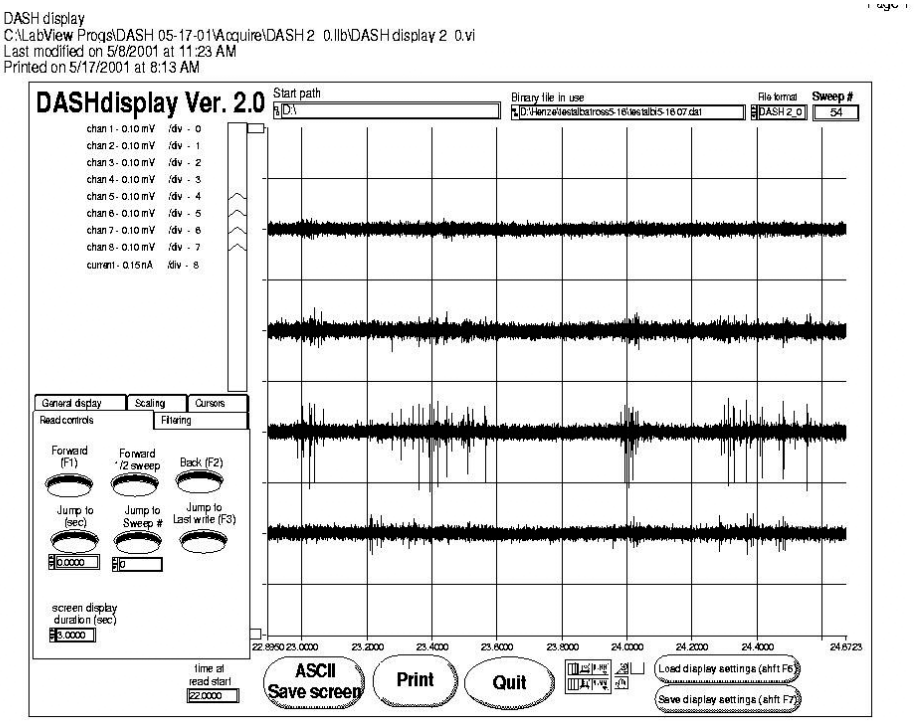

Fig. 14: Filtered (100Hz-1kHz) data from the previous traces, clearly showing single unit activity, especially on channel three.

On-chip buffering is important in that it provides the possibility of decreased coupled noise. In behavioral experiments in particular, the group at Rutgers has found that motion artifacts can significantly degrade the signal by obscuring spikes and field activity, and in some cases even saturating the amplifier. Since coupled noise will be most significant at high impedance nodes, it is postulated that significant noise is being coupled into the electrode before the discrete unity-gain headstage amplifier. If this is the case, an integrated buffer stage should improve the situation. In fact, initial experiments indicate that the motion artifact is improved, and more rigorous tests are underway. There is an additional argument for on-chip buffering, which is that discrete headstage amplifiers are bulky and difficult to assemble, becoming prohibitive at very high channel counts. For purposes of illustration, a rat with a 96-channel passive probe implanted chronically is shown in Fig. 15. Three 32-channel headstage amplifiers, enlarged in inset, are connected during a recording session. It is worth noting that even with this somewhat cumbersome set-up, useful data has been generated. In fact, the probe and data recorded in rat hippocampus was featured in the July 17 issue of Nature (Fig. 16). However, the setup is clearly cumbersome and impedes the movement of the animal, which is supposed to be free to move in its environment. In addition, it has been difficult to obtain useful recordings during movement due to coupled noise. In order to solve these problems (and as previously reported), a fully integrated 96-channel buffered probe has been fabricated. Initial experiments have recorded field and unit activity; however, these experiments indicate that the active probes are very optically sensitive. This is not unexpected, due to the presence of a parasitic reverse-biased diode caused by the source/drain regions of the transistors in the analog selectors. In order to address the problem, previously fabricated wafers are being completed with an electroplated gold circuit shield.

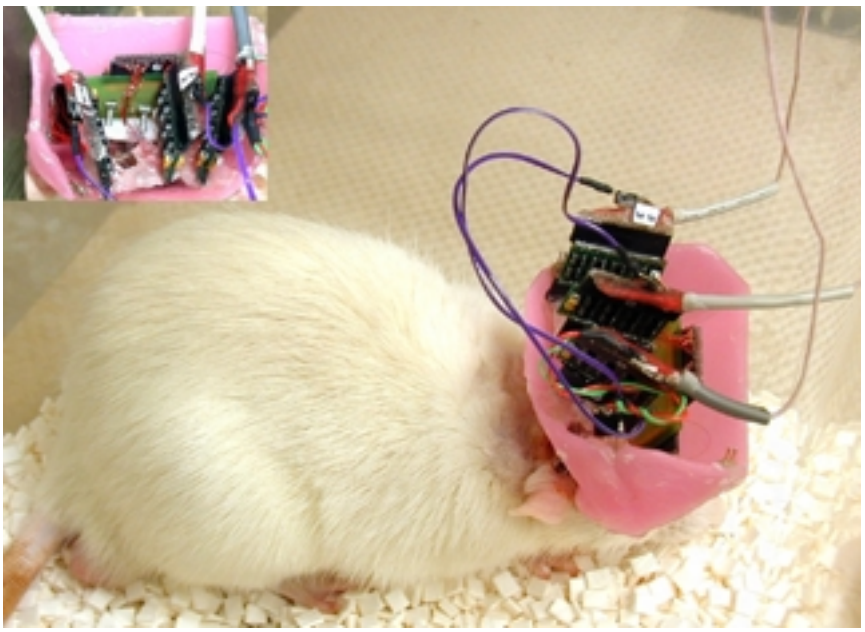

Fig. 15: Photo of rat with chronically implanted 96-site probe. Discrete stage amplifiers (see inset) are cumbersome, and motion-induced coupled noise is excessive.

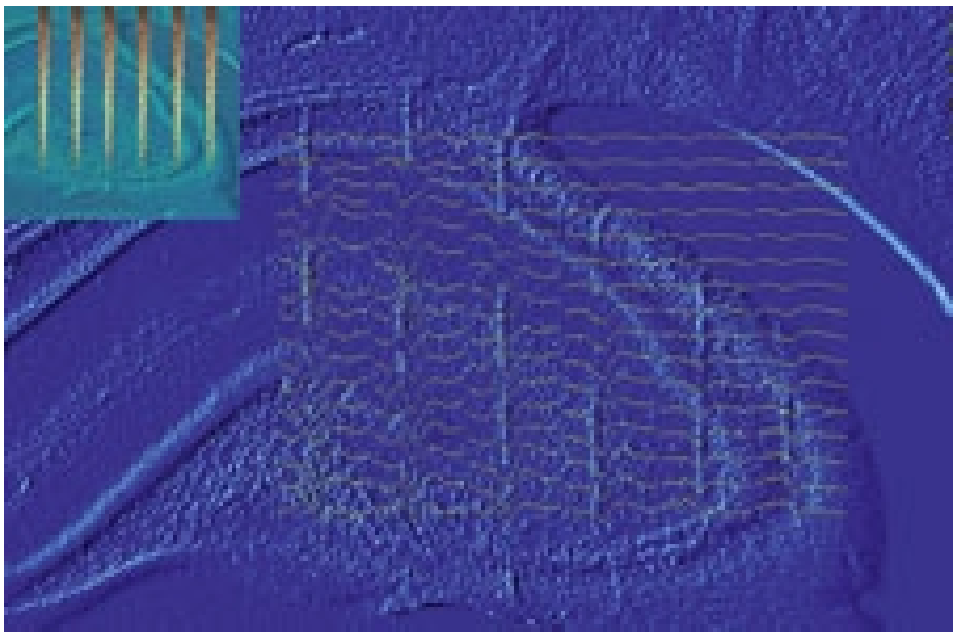

Fig. 16: A figure (which appeared in *Nature*, July 17, 2001) showing data recorded from the passive 96-channel system (in an immobilized rat) and showing the probe tips (inset).

The experiments at Rutgers also indicate that the DC offset at the output of the probe may cause trouble with the input of certain commercial amplifiers. Although the wide-field data is high-pass filtered at 1Hz, it was found that one of the two amplifiers in the lab saturated when connected to the probe. Because the on-chip buffer is a source follower, there is a DC shift in the output level, and it is hypothesized that this DC offset (typically 1.5 to 2 V.) is saturating the amplifier. This problem can be addressed by going to a common-source configuration, which means that the output node of the driver is a drain, and can thus be independently biased. There are other problems with common source stages, most notably that they are sensitive to input DC level and that they depend on the output impedance of the load for gain. Thus, it is difficult to bias the transistor in the saturation region while accounting for the DC drift of the electrode. In addition, in order to achieve even unity gain with a reasonably low output impedance, both the input and output transistors must be made very large. Still, a probe utilizing common source amplifiers with a resistive load has been designed for fabrication on the upcoming active run. If successful it does offer some advantages. The probe will have a lower output impedance than the source follower (500Ω) without the need for a large load transistor. In addition, because the bias string of the source follower contributes to 1/f noise, the common source configuration (which has no bias string) exhibits significantly less noise in bandwidth. Finally, because the load is resistive, the output can swing relatively close to ground (several hundred millivolts) which would not be possible with a transistor load. This will allow us to set the DC output close to ground and hopefully solve the problem of the saturating amplifier.

A more elegant solution to these problems is in the use of an operational amplifier in unity-gain configuration. An op-amp suitable for neural recording has been fabricated and reported in previous quarterly reports. This amplifier has been used in closed-loop configuration with a pass-band gain of 100, and it is AC-coupled with a lower corner frequency of approximately 10Hz. In this configuration, it is suitable for single unit recordings; however, to record both spikes and field data several modifications need to be made. The capacitive feedback scheme is not ideal because the lower pole must be made close to 1Hz. This is difficult to achieve while allowing for process variations. A better solution may be to short the output to the inverting terminal, putting it in a follower configuration. This configuration has the added advantage that it allows the measurement of the DC site potential simultaneously with the field and unit data. Because this configuration is less stable, it was necessary to re-compensate the amplifier. In addition, the increased decade of bandwidth placed a more severe constraint on transistor noise, which is dominated at low frequency by 1/f noise. This necessitated significantly resizing the input transistors in order to increase gate area and decrease 1/f noise. Probes with the unity gain configuration op-amp have been included on the current active run for use in Buzsaki's laboratory (Fig. 17).

Finally, more work is needed to explore the chronic recording capability of the active probes. The 96-site experiments are chronic, and the full advantages of the active probes are most apparent in this type of preparation. This is uncharted territory, and a series of specially designed chronic probes with test structures have been designed and included on the upcoming fabrication run (Fig. 17). It is hoped that the knowledge that we gain from these probes (for example on passivation layers and coatings, bias levels and the effect on tissue, etc) can be passed along for use in the experimental work at Rutgers.

7. A 64 Site Neural Recording Probe with On-Chip Amplification and Multiplexing (PIA-2 and PIA-3)

Four new acute 2D neural recording arrays, each featuring different amplifier configurations, have been designed and are ready for fabrication. The probes have sixty-four sites of which eight can be selected for amplification. The sites are spaced 100µm apart vertically and 200µm apart horizontally. The first stage of amplifiers provides an in band gain of 40dB while filtering signals above 10kHz as well as any dc baseline polarization of the electrode. The outputs of the eight amplifiers are next time-division multiplexed onto one line and passed through an active filter. The filter provides an additional signal gain of 20dB while attenuating clock noise coupled into the signal path by the time-division multiplexer. The probes feature multiple test modes that allow each of the components to be individually characterized. The layout of a 2D neural recording probe is shown in Fig. 18 below. The back end of the probe is 1.86mm by 3.2mm.

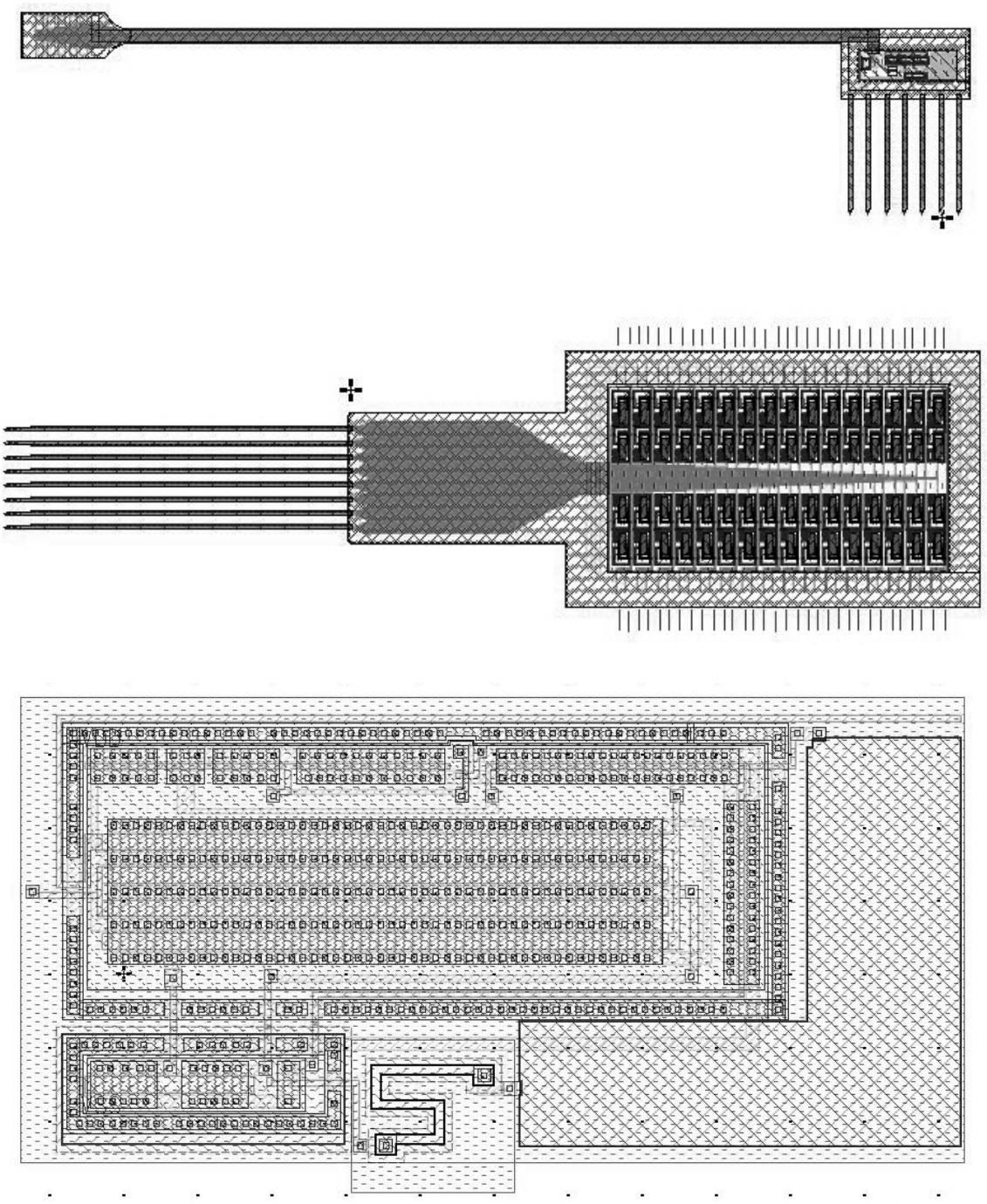

Fig. 17: Two of the recording probes included on current active fabrication run. At top is an active probe with integrated ribbon cable. The circuitry includes test structures to monitor the etching of the top dielectrics and measure threshold shifts, as well as buffering and select circuitry for chronic in-vivo recording. Middle, a 64-site buffered probe which incorporates the op-amp in unity-gain configuration (bottom.)

Two of the circuit configurations featured on the acute probes have been included on arrays for 3D assembly. Due to the reduced number of data leads, the testability included on the acute arrays could not be utilized in the 3D designs. The layout of a chronic recording array is shown in Fig. 19. The back end of the probe is 1.9mm by 3.5mm.

A Probe for Telemetry Applications

The on-platform A/D converter was implemented as a fully differential design to maximize the dynamic range. Therefore, a single-ended-to-differential amplifier stage was designed to interface between the standard single stage op-amp and the A/D converter as pictured in Fig. 20. Schematics of the differential amplifier, its common mode feedback circuit, and its kt/q reference are shown in Figs. 21, 22 and 23, respectively. A probe level layout of the differential amplifier is pictured in Fig. 24. The amplifier features two class AB output stages for wide output swing and low quiescent current. The common mode feedback circuit is functional over a wide differential input range. Thus, the amplifier utilizes most the input dynamic range of the on-platform oversampling A/D converter. Table 1 summarizes the amplifier performance.

Gain	31.5dB
Bandwidth	1Hz to 10KHz
Power Consumption	329μW
Layout Area	0.243mm^2
Output Swing from ±1.5V Supply	±1V

Table 1: Differential Amplifier Specifications

Future implementations should move the differential amplifier from the probes to the platform for the following reasons. First, switched-capacitor common-mode feedback circuits can achieve a large differential input range without the use of resistors. In its present continuous-time implementation, the resistors in the common-mode feedback circuit reduce the open-loop gain of the op amp by loading the output. This prevents the use of multiple common mode feedback stages because the reduction in forward gain due to the resistors is unacceptable. In order to achieve higher output swing, a class AB output stage such as the one pictured in Fig. 25 can be used. This, however, adds an extra pole in the common mode feedback loop that causes stability problems. This can be remedied by the addition of more common mode feedback circuits. As mentioned earlier, this solution is not viable for continuous time implementations. Thus, to be able to take advantage of the entire input dynamic range of the on platform A/D converter, the differential amp should also be included on the platform so that switched capacitor common mode feedback circuits can be used in conjunction with wide swing class AB output stages.

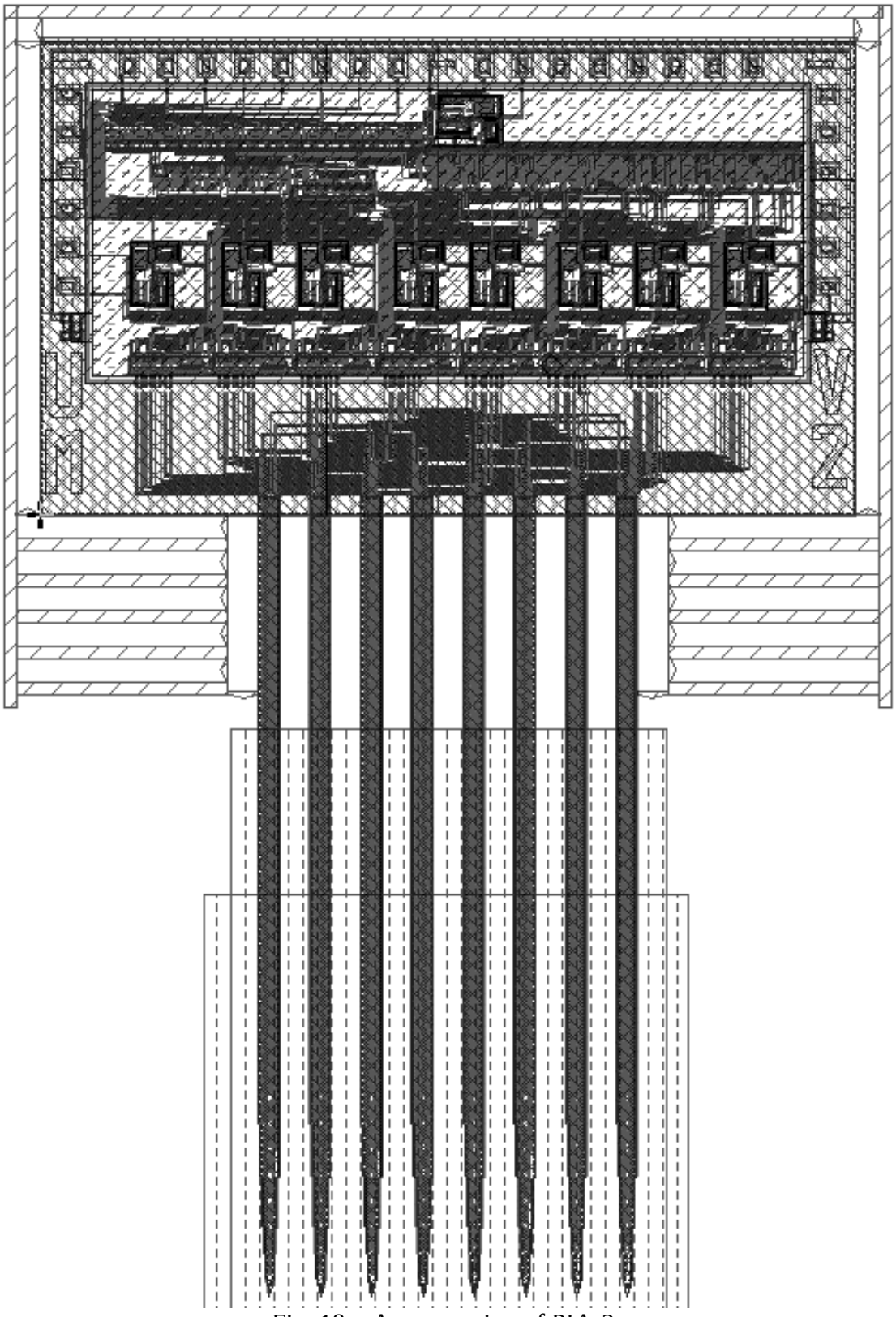

Fig. 18: Acute version of PIA-2

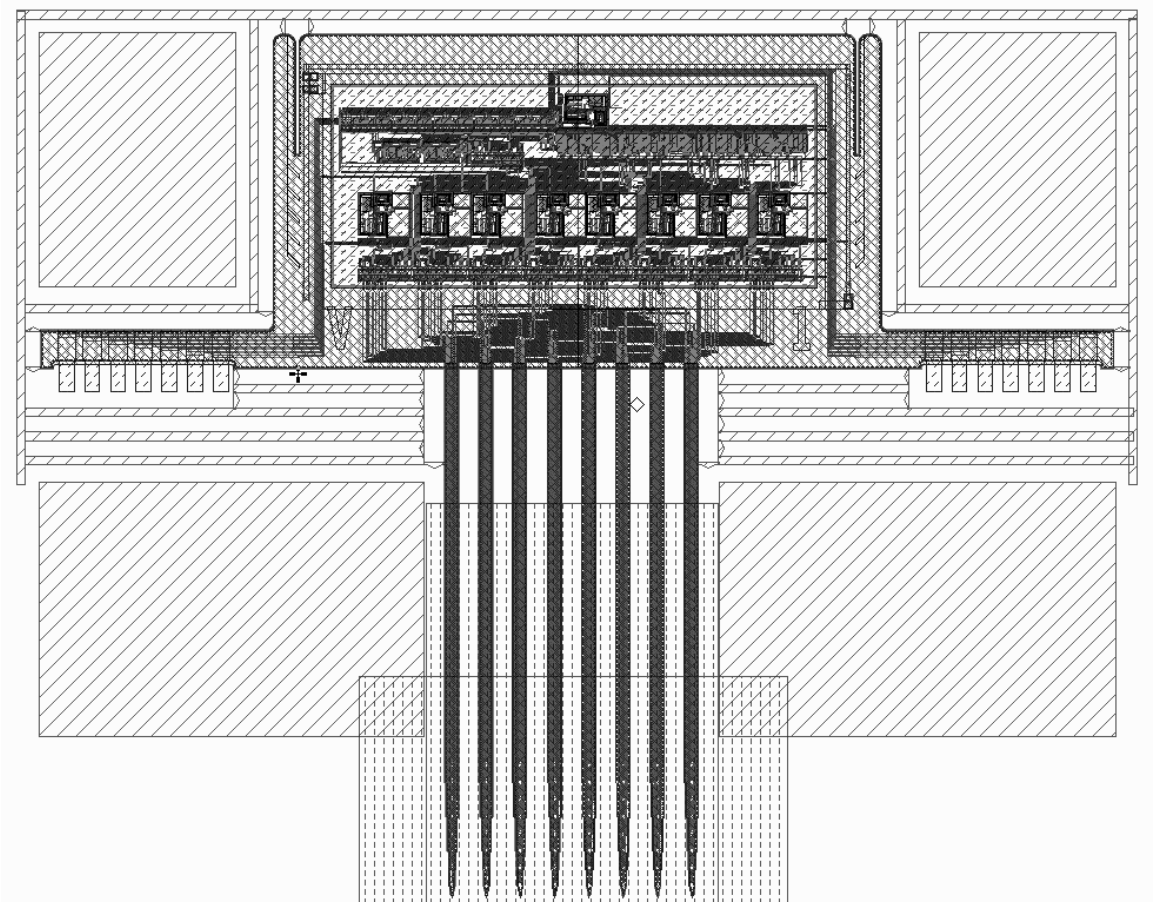

Fig. 19: Chronic recording probe

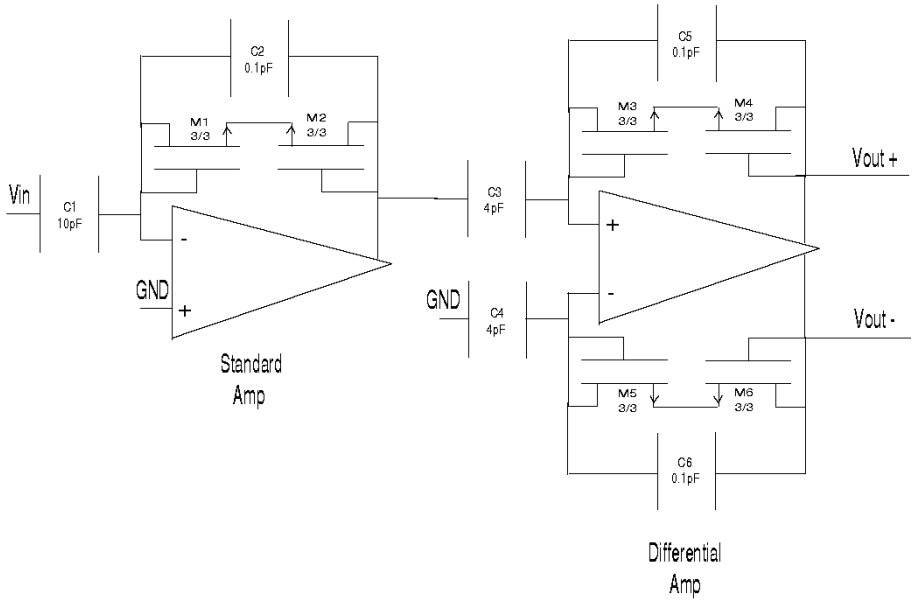

Fig. 20: Telemetry amplifier

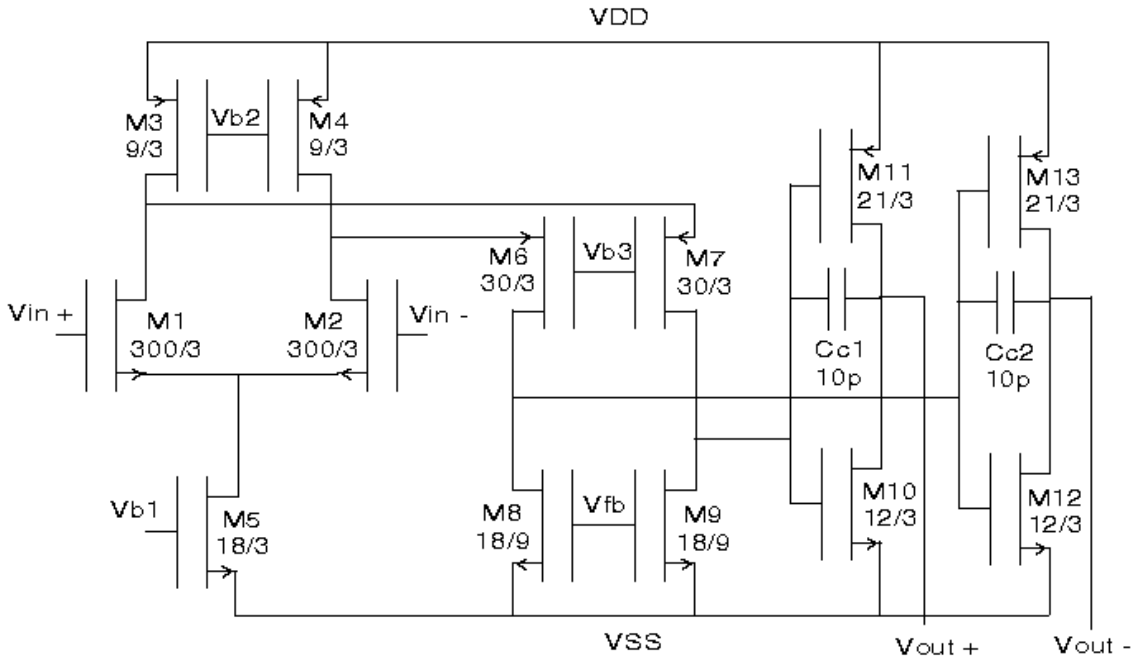

Fig. 21: Differential amplifier

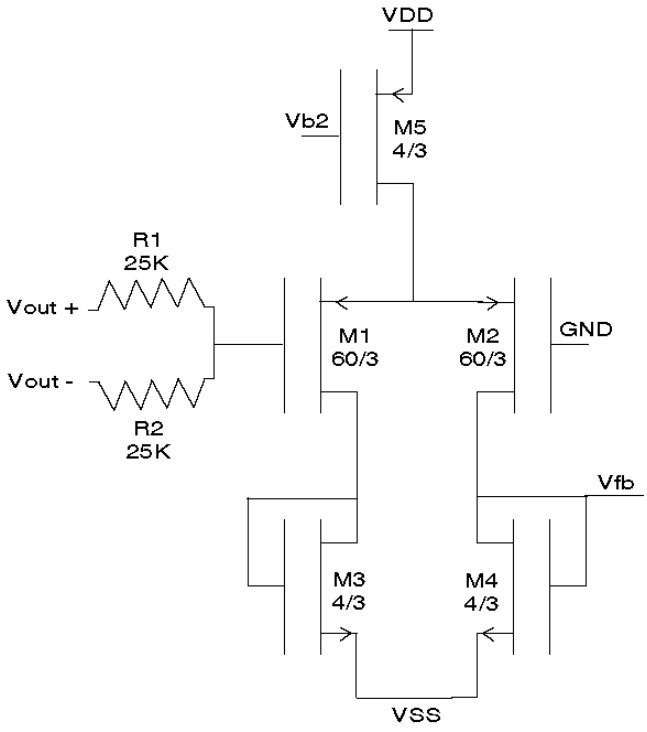

Fig. 22: Wide-swing common-mode feedback circuit

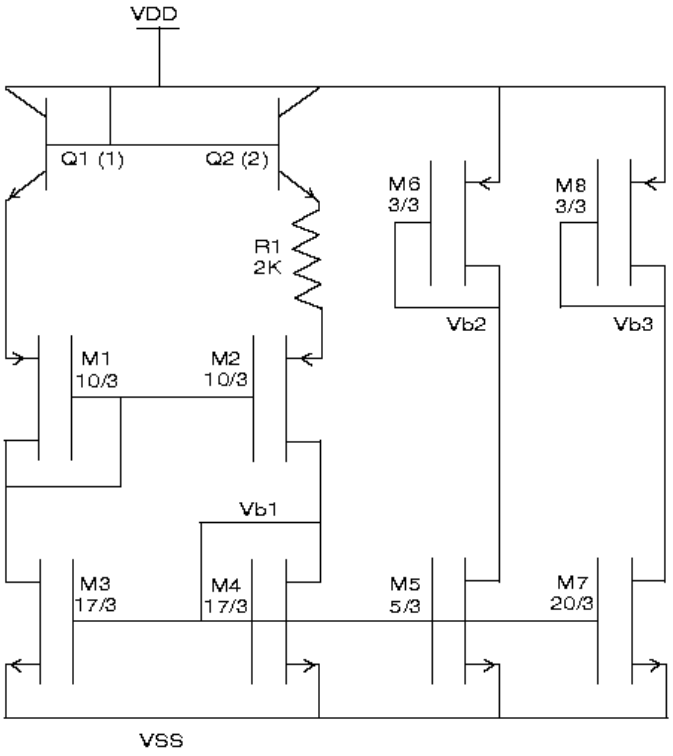

Fig. 23: kT/q reference

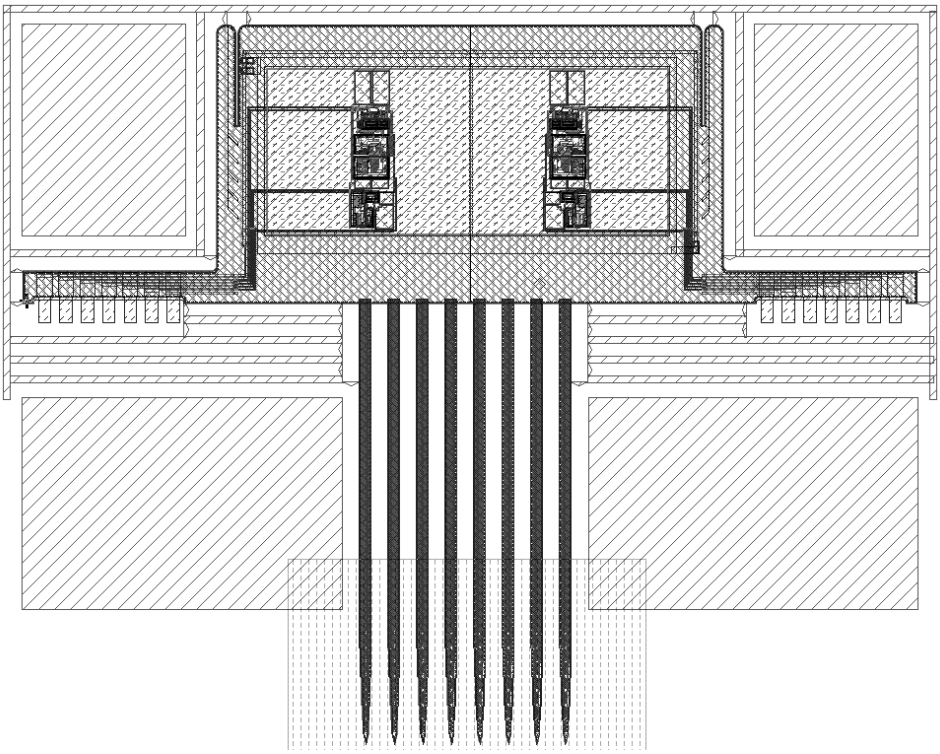

Fig. 24: Telemetry probe

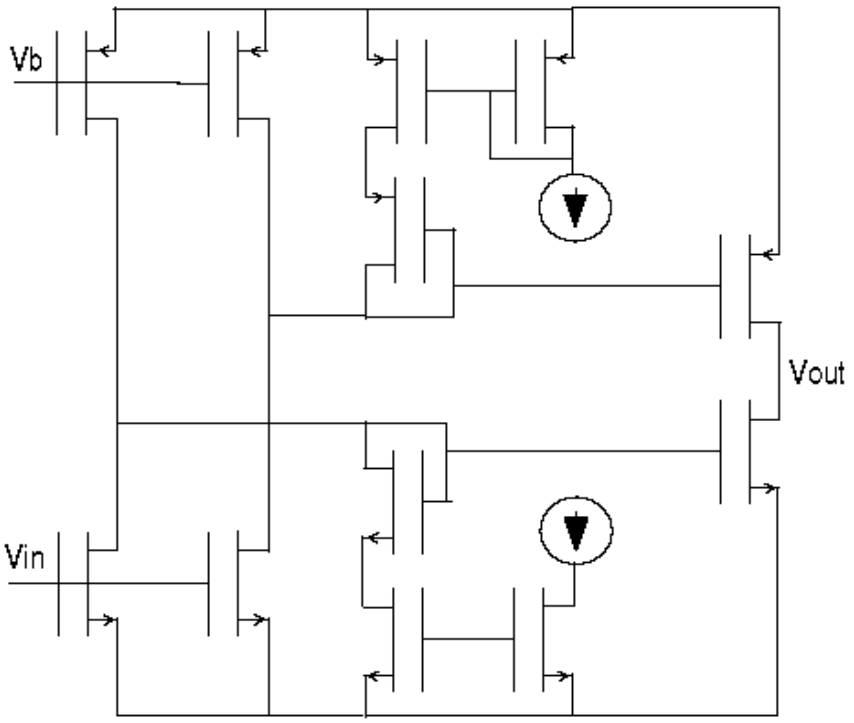

Fig. 25: Wide-swing output buffer

8. Design of a Wireless Telemetry Platform for Multichannel Microprobes

Design of the on-chip transmitter

In the development of an implantable wireless interface for the probes during the past quarter, an on-chip transmitter was designed to send the digitized recording data out to the external world. A challenge still exists in lowering the power consumption. Figure 26 shows a schematic of this transmitter.

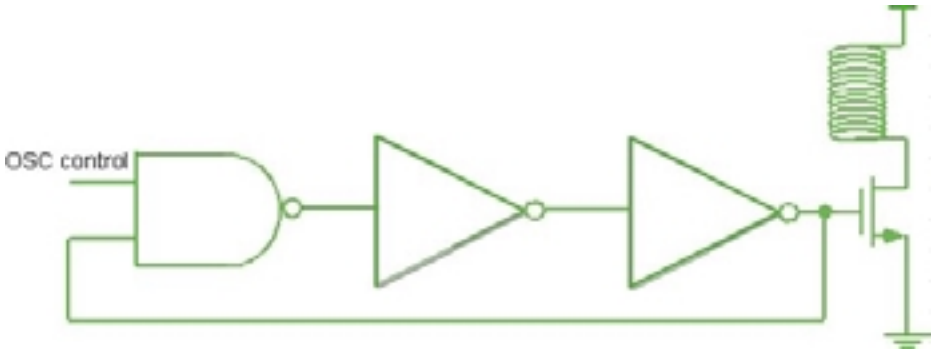

Fig. 26: The schematic of the on chip transmitter

As shown in Fig. 26, the control signal OSC comes from the output of the sigma-delta modulator, which determines the oscillation of the ring oscillator. Using this method, OOK (on-off-keying) is used for the data communication. In the above design, the frequency of the oscillator can reach more than 100MHz, based on the simulation.

Layout of the telemetry interface

The design of the telemetry interface for one-channel of neural recording had been finished and the design procedure and the simulation results have been shown in previous reports. The entire telemetry used for one-channel neural recording was laid out and submitted to MOSIS for fabrication. The layout of the chip is shown in Fig. 27.

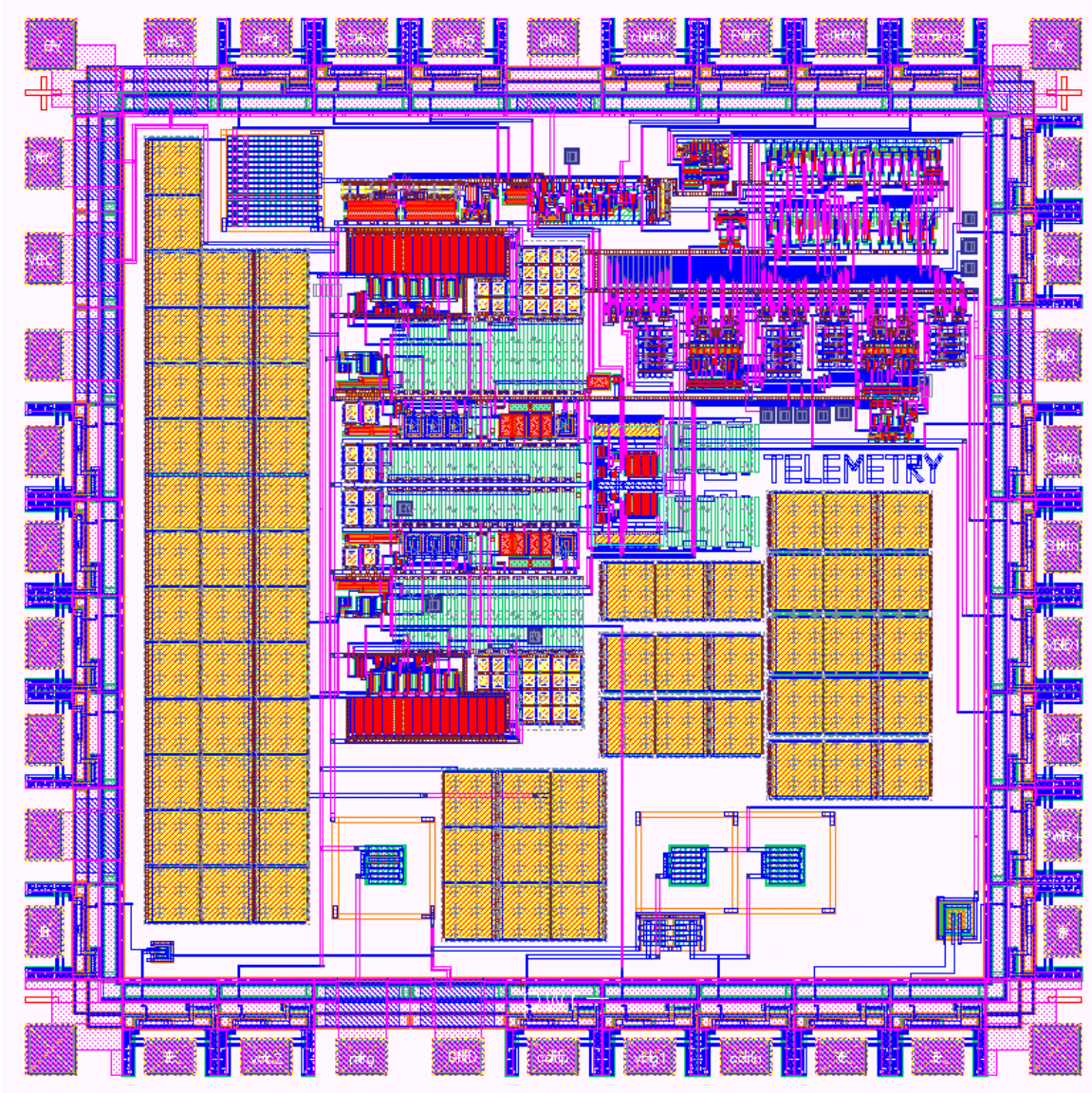

Fig. 27: The layout of the telemetry interface for one-channel of neural recording

This layout was done using Cadence, the circuit netlist was extracted from the layout, and post-layout simulations were carried out to verify functionality. A layout summary is listed in Table 2.

Processing	AMI 1.5um, Standard CMOS processing
Area	2.2x2.2mm^2
Package	*DIP40*
Circuits on chip	Front-end, Sigma-Delta, Transmitter, Control Logic

Table 2. The summary of the layout of telemetry chip

A summary of circuit die area and power consumption by circuitry block is shown in Table 3.

Circuit Blocks	Power Consumption	Area(um^2)
Two Regulators V=3.3v *One for Digital* One for Analog	2 x 70µW	2 x 500 x 500
A Regulator V=1.65v Used in DSM and Preamplifier	15µW	160 x 100
Clock Recovery	50µW	120 x 100
POR + Voltage limiter	< 3µW after Reset	60 x 100 + 80 x 100
Ask Demodulator	20µW	350 x 100
Total Front- end Circuitry	230µW	0.33mm^2
Two Integrators in DS_	2 x 50µW	2 x 125 x 170
Clock generator for DSM	70µW	370 x 250
1 bit ADC + CMFB	8µW (working at 2MHz)	70 x 100 + 150 x 80
Total for DSM	200µW	0.42 mm^2
Preamplifier	100µW	----
Manchester Decoder	5µW at 200kHz Command Bit rate	80 x 100
Total for the chip	<1mW	2.2 x 2.2 mm^2

Table 3: The summary of the power consumption and area of the telemetry chip

Design of the external transmitter

In order to supply enough power to the receiver coil, a high efficiency transmitter/amplifier has to be used, where a Class-E power amplifier is a good candidate. A simplified class-E power amplifier is shown in Fig. 28.

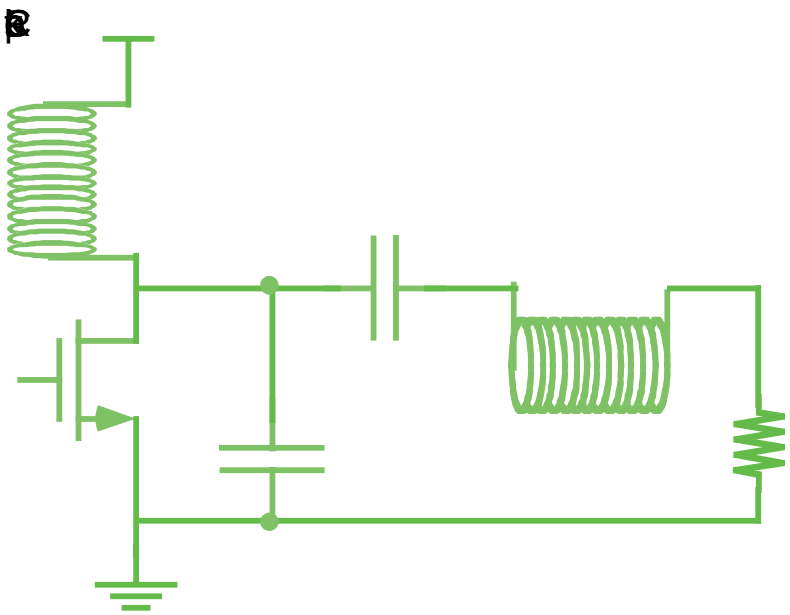

Fig. 28: A simplified class-E power amplifier

Referring to Fig. 28, a high-speed power switch drives a resonant network consisting of a parallel capacitor (Cp), a transmitter coil (Lt), a tuning capacitor (Ct) and a resistive load (Rs). When the switch is off, the choke inductor (L_{choke}) acts as a current source charging the resonant network and creating a transient voltage across the switch. When the switch is on, its current rises smoothly until the switch is off again. Losses are kept at minimum by having the transistor switch on when both voltage and current are small. The current and voltage across the transmitter coil are sinusoidal at the resonant frequency of the tank circuit, which has to be the same as the driver switching frequency. The efficiency of Class-E transmitter can reach 70-100% with proper design.

The complete class-E transmitter includes four major blocks: 1) the class-E amplifier (switching transistor and resonant tank), 2) the feedback circuitry, 3) the start-up circuitry, and 4) the amplitude modulator. The schematic of the complete class-E transmitter is shown in Fig. 4. The Class-E amplifier has been addressed in previous paragraphs.

In the feedback circuitry, the zero-crossings of the delayed tuning capacitor voltage are used as a time base to drive the switching transistor. By introducing a 90 degree phase delay, the class-E mode of operation can be ensured regardless of any inadvertent variations of the transmitter coil inductance.

A start-up circuit was designed to start the oscillations when the transmitter is switched on. The amplitude modulator utilizes the principle that the voltage across the inductor in the series load is a linear function of the supply voltage. By adjusting R1, the degree of modulation is easily varied to accommodate the desired levels.

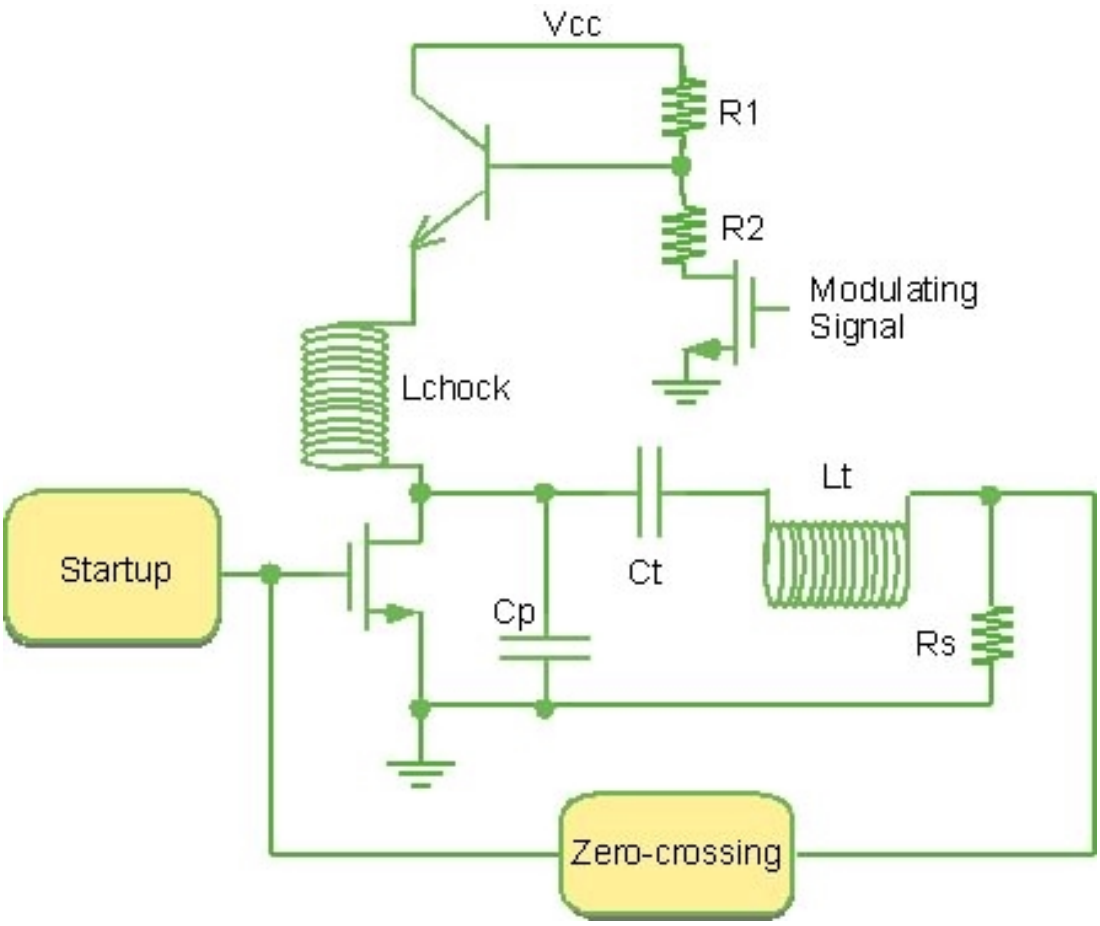

Fig. 29: The schematic of the external transmitter

The external transmitter circuit is still being developed; the detailed simulation results will be reported in the next quarterly report.

9. Conclusions

During the past quarter, we have continued to explore the long-term recording characteristics of these probes in-vivo. One 8-shank probe having both large tip sites and standard center-shank-mounted sites is still recording well from guinea pig auditory cortex after 249 days. We have also explored the use of rough gold plating and found it effective in reducing site impedances as well as in providing increased adhesion for electrochemically-deposited conducting site polymers such as polypyrrole doped with polystyenensulfonate (PPy/PSS).

A new high-density connector has been built using a polyimide-based flex circuit rather than the traditional rigid FR-4 board. The flex circuit is double-sided which

permits two dual-row Omnetics connectors to be used to achieve 32 channels in a package measuring 12mm tall x 9mm wide x 5mm thick. Implants using these packages will begin during the coming quarter. Also during the past term, we have begun to re-examine the possibility of reducing probe shank width to dimensions below 10µm. This should result in significantly-reduced tissue damage on insertion and permit lateral recording between shanks within the recording field of single cells. We will explore the fundamental limits on shank width as well as the use of alternative technologies such as silicon-on-insulator to achieve greater compatibility with foundry fabrication of the probes.

A complete graphical user interface for the non-multiplexed 64-site 8-channel PIA-2B probe has been developed, and work with this probe is proceeding in collaboration with Gyorgy Buzsaki's laboratory at Rutgers University. The probe has reduced recording artifacts and been used to drive ten-foot cables without a headstage. Results at Rutgers using the passive version of a companion 96-site probe in rat hippocampus were reported in the July 17 issue of *Nature*. We have also designed a version of the 96-site probe using op-amp-based buffer amplifiers and look forward to using these devices during the coming term.

Several versions of multiplexed 64-site 8-channel active probes (PIA-2 and PIA-3) have been designed and are now in fabrication. The probes feature different amplifier designs and a variety of test modes. In addition, we have designed a probe for use in a completely wireless recording system. The probe features an on-chip gain of 31.5dB, a bandwidth from 1Hz to 10kHz, a power dissipation of 0.32mW, a layout area of $0.24mm^2$, and a swing of ±1V from a ±1.5V supply. A platform-mounted telemetry interface for the probe has been designed and submitted to MOSIS for fabrication. The interface dissipates less than 1mW in a layout area of 2.2mm x 2.2mm. We hope to demonstrate completely-implantable wireless neural recording before the end of the year.